Towards Fluidic Actuation for Catheter-Based Interventions

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Abstract:
The minimization of surgical procedure invasiveness poses considerable challenges. During minimally invasive surgeries (MIS) surgeons encounter two main problems: a lack of visualization and reduced control over the instruments. In order to assist the surgeons in their task, several robotic instruments have been developed. In fact robotic catheters could increase the success rate and reduce the number and severity of complications in catheter-based interventions such as trans-catheter aortic valve implantations (TAVI). This abstract introduces a novel inherently compliant active catheter using fluidic actuation. The large compliance of the proposed catheter is beneficial compared to catheters that are steered by more traditional cable-based actuation. The catheter configuration and the design process are introduced in this paper. Finally, a steering control is proposed.

Keywords: Active Catheter, Fluidic muscle, Catheter Control

Introduction

The invasiveness of surgical procedures has always been a major concern of patients and the surgical community. Minimally invasive surgery techniques originated from this concern. Although advantages of MIS for the patient such as e.g. fast recovery are undeniable, surgeons are facing substantial increases in complexity. The two main problems they encounter are a lack of visualization and a lack of control over their instruments. While the current therapeutic trend towards MIS poses considerable technical challenges, it also forms a catalyst for the development of new robotic technology that is designed to overcome these complications. Depending on the procedure, different robots have been developed in the past in order to assist the surgeons in their task. Often this happened at the expense of an increase in operation time and instrumentation footprint [1].

Robotic catheters for trans-catheter aortic valve implantation (TAVI)

Patients with severe aortic stenosis, previously classified as inoperable can now be operated using a trans-catheter approach. During current TAVI procedures the catheters such as the NovaFlex+ Transfemoral System (by Edwards Lifescience) used as valve delivery system only provide a single manually controllable bending motion. Because of the stiffness and the poor controllability of these delivery systems, their use goes not without risk of puncture or tissue damage. Rather than directly inserting catheters through the groin to reach the aortic valve, first a flexible hooked guide-wire is inserted and advanced with the help of X-ray imaging. Once the heart's left chamber has been reached, the valve delivery system is slid over the guide-wire and brought up to the aortic annulus.
In order to reduce the complexity and provide the surgeons a better intra-vascular mobility, robotically actuated catheters could be used and, potentially enhance the success rate and reduce the number and severity of complications [2]. However, current robotic solutions are often either too big or too stiff to work in such delicate, moving and deformable environment. Flexible continuum robots provide an appealing alternative [3].

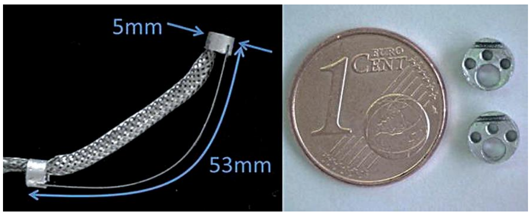

Fig. 1: *(left) Unidirectional bending segment without protection sleeve. With protection sleeve the outer diameter rises from 5 to 6mm; (right) 5mm diameter connection parts, including three 1.1mm working channels.*

A novel fluidic actuated catheter

This paper presents a novel robotic catheter. Fluidic actuation is used to yield a catheter that is inherently compliant and thus safer for the patient. The proposed continuum robot is made of multiple unidirectional bending segments connected in series, each of which being independently controllable. Three additional lumens were foreseen here to allow integration of sensors or passage of instruments.
In the remaining part of the paper the actuation principle as well as the bending structure are

described. The characteristics of the unidirectional bending segment and the developed novel catheter are presented next. Finally, some conclusions and future work are listed.

Fluidic Actuation

Each bending segment is actuated by a pneumatic artificial muscle (PAM) and provides one controllable degree of freedom. PAM actuators are well known for their high power density and have already been extensively described, modelled and used in the literature [4] [5]. Most of the developed models rely on an energy balance or force balance and agree on the major force component, also known as the Gaylord force, which is described by:

$$F_G = \frac{P}{4N^2\pi}(3L^2 - B^2) \qquad (1)$$

where F_G is the contracting force, P the actuation pressure, L the actuator current length, B the length of one fibre of a braided sleeve that forms the external part of the actuator and N the number of turns a fibre makes around internal rubber tube of the muscle [6].

When the muscle is rigidly clamped between a pair of yokes – as depicted in Fig. **2** – and pressure is increased the Gaylord model predicts a linear rise in contracting force (blue dashed line in Fig. **3**). The figure also shows the force that was actually measured during (in solid red line) an expansion and relaxation cycle. From the figure it can be seen clearly that the Gaylord model overestimates the real force. More detailed models that capture also the non-linearities such as amongst others the elasticity in the braid and bladder, end effects as well as friction have been proposed in the scientific community [5]. These models can be used for advanced control of the catheter.

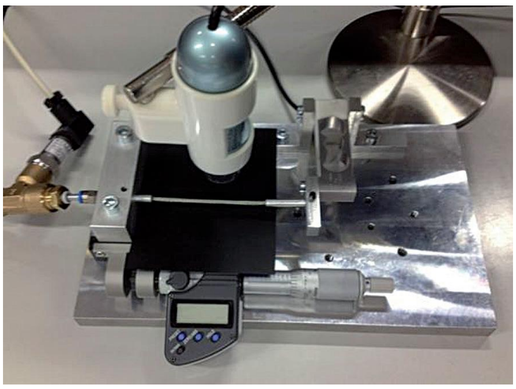

Fig. 2: *PAM test bed using a pressure sensor, a load cell, a micrometer screw, and a microscope.*

The artificial muscles developed here have 2.2mm outer diameter at rest and can reach a contracting stroke up to 25%. The test setup presented in Fig. **2** was used to measure the evolution of the blocked force with the applied pressure. The blocked force is defined as the contracting force produced by the actuator for a fixed length. As can be seen from Fig. **3**, the maximal force reached at 6.35 bars is equal to 12.6 N for a 70mm long muscle.

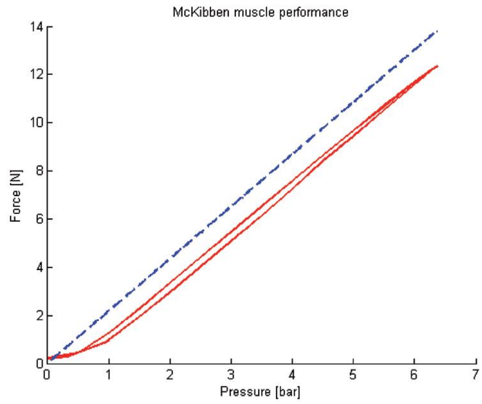

Fig. 3: *Blocked force vs internal pressure for a 70mm long PAM with an external diameter at rest of 2.2mm (solid line); Gaylord force (dashed line).*

Bending structure of catheter

In order to transform the linear contraction of the artificial muscle into a bending motion, each segment of the catheter is build up as an asymmetric structure. A leaf spring is placed in parallel to and at a distance of 1.7 mm from the muscle and ensures this behaviour by constraining the segment length due to its axial stiffness (see Figure 1 (left)). The spring that was used is 0.2mm thick, 3mm wide and has a Young's modulus of 186GPa.
In each unidirectional bending segment the pneumatic artificial muscle and the leaf spring are attached to two connection parts of 5mm diameter. They were also manufactured in order to provide three additional lumens to integrate sensors. These lumens could also be used as a tool channel through the catheter since they all have a diameter of 1.1mm. Figure 1 shows a 53mm long bending segment without its protection sleeve actuated at a pressure of 6.35bars as well as the manufactured connection parts.

Around the previously described bending structure an outer braided sleeve is placed in order to constrain the distance between the muscle and the leaf spring while the structure is bending. This allows a higher bending while decreasing the required space. With its outer sleeve, the total diameter of the catheter prototype rises up to 6mm.

As shown in Figure 4 when the pressure increases from 0 to 6.35 bars, the bending angle varies from 28° to 130°. The maximal pushing force delivered by the bending segment has been measured as follows. The unactuated segment was positioned parallel to a single point load cell (Model 1668(S), BCM Sensor Technologies) (see figure 5(left)), and then actuated with 6.35 bars. The resulting shape of the bending segment is shown in **Fig. 5** (right). The maximal force measured is 0.61N. The large difference between the maximal blocked force which is larger than 12N and the measured pushing force of 0.61N is mainly explainable by the contracting bending moment that is needed to deform the asymmetric structure. After this deformation only a relatively low force remains to push on to the load cell.

Fig. 4: *70mm long bending segment actuated with 6.35bars (left), and not actuated (right).*

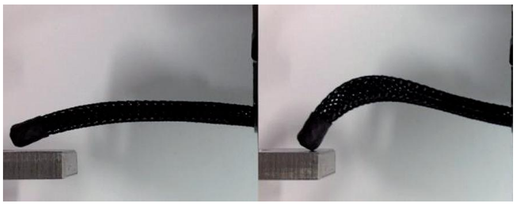

Fig. 5: *Force measurement of fluidic muscle*

In order to compare the compliancy of the presented unidirectional bending segment with the currently used NovaFlex+ Transfemoral System, an experiment has been set up to measure the respective stiffness. The measurement system is similar to the one presented in Figure 2. The catheters were positioned parallel to the load cell. The force and the deflection of the 70mm long bending segment and the NovaFlex+ were measured simultaneously in their natural bending plane. In both cases contact was made at a distance of 70mm from the clamping point. The results of this compliance experiment are presented in **Fig. 6**. Similar measurements have been done where the force and deflection were measured perpendicularly to their natural bending plane. In that case, the stiffness of the developed bending segment was similar to the stiffness of the NovaFlex+. The stiffness of the latter has also been measured at different distances from its tip. Limited differences were observed here. The average stiffness of the developed bending segment is 9.8[N/m] which is approximately 12 times smaller than 119[N/m] for the NovaFlex+.

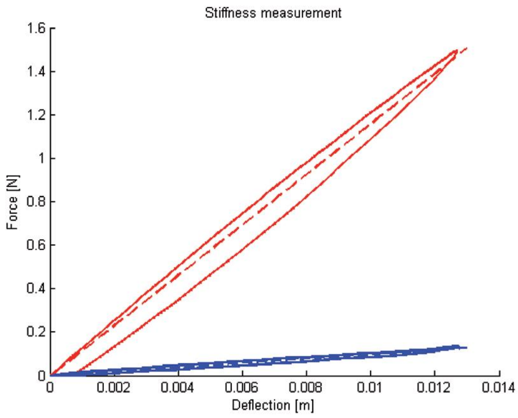

Fig. 6: *Stiffness measurements for the developed catheter (solid blue line) and for the NovaFlex+ (dotted red line). A first order fitting is plotted in dashed lines for both cases.*

Novel robotic catheter

The developed catheter is composed of two unidirectional bending segments connected in series (see Figure 8). The distal segment has a length of 70mm and the proximal one of 30mm. Electromagnetic position sensors have been integrated in the foreseen lumens in order to provide shape sensing information. The position sensors used are Aurora sensors developed by NDI Medical. These sensors typically provide position accuracy better than 1mm within their working range. The entire workspace of the presented catheter prototype has been measured using these sensors and is presented in Figure 7. Two 5DOFs sensors were respectively placed at the base of the proximal bending segment (cross) and at the tip of the distal bending segment (lines). During the measurements the catheter was clamped at the level of the proximal sensor. The two actuators were randomly pressurized over their 0 to 6.35bar operating range.

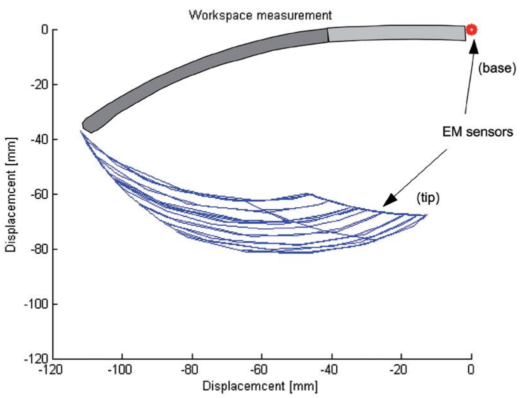

Fig. 7: *Workspace measurement – in grey: conceptual sketch of 2DOF actuator – in blue: workspace as it is measured by the tip-mounted EM sensor.*

As it can be observed from Figure 7, for the purpose of clarity, the two bending segments are both bending in the same plane and direction. But in order to provide different intra-vascular mobility, the different actuated segments can be, before the assembly procedure, axially rotated with respect to each other. The general configuration of the catheter as well as the length of each bending segment can be optimized using a patient-specific catheter design approach described in previous work by the authors [7] where the developed workspace is mapped to the required operating workspace.

Fig. 8: *Robotic catheter with 2 segments navigates through a silicon aorta phantom.*

Conclusion and Future Work

A novel active catheter using fluidic actuation has been developed. The new catheter is a factor more compliant than currently used cable-driven catheters. The novel catheter also provides a large stroke. Shape sensing can be easily integrated in this type of catheters.

Based on pre-operative data the catheter configuration can be optimized in order to provide an optimal intra-abdominal mobility. Combining shape sensing information provided by position sensors with pre- or intra-operative data, the robotic catheter can be steered along a strategic trajectory. For example it could be programmed to try and follow the aorta centreline, minimizing the interaction force between the catheter and its environment. Due to its compliant nature even if specific trajectories cannot be followed the interaction force and possible damage following are expected to be substantially lower than is the case for more traditional design.

Further characterization and miniaturization of the bending segments remain topics for future research.

Acknowledgments

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